Regional wave travel and reflections along the human aorta: a study with six simultaneous micromanometric pressures

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ABSTRACT The human aorta and its terminal branches were investigated in normal subjects during elective cardiac catheterization to evaluate regional wave travel and arterial wave reflections. A specially designed catheter with six micromanometers equally spaced at 10 cm intervals was positioned with the tip sensor in the distal external iliac artery and the proximal sensor in the aortic arch. Simultaneous pressures were obtained and analyzed for foot-to-foot wave velocity, and Fourier analysis was used to derive apparent phase velocity. These quantities were assessed during control (n = 9), during Valsalva (n = 8) and Müller (n = 4) maneuvers, and during femoral artery occlusion by bilateral manual compression (n = 8). During control, regional cross-sectional areas, determined from aortography, and regional foot-to-foot pulse wave velocities were used to calculate the local reflection coefficient in the proximal descending aorta (Γ = 0.05), at the junction of the renal arteries (Γ = 0.43), and at the terminal aortic bifurcation (Γ = 0.13). To test the hypothesis that significant reflections originate in the aorta, at the level of the renal arteries, aortograms were used to design a latex tube model with geometric properties similar to the descending aorta. Velocities and reflection characteristics in the model and in vivo were compared. Inspection of thoracic aortic pressures under control conditions revealed a reflected wave originating from the region of the aorta at the level of the renal arterial branches while abdominal pressures exhibited reflection from a site peripheral to the terminal aortic bifurcation. In the low frequency range, apparent phase velocity was found to be higher proximal to the renal arteries as compared with at the distal sites. In addition, the minimum value occurred at a higher frequency in the lower thoracic aorta than at more distal sites. The effects of reflection on apparent wave velocity in the tube model were consistent with data obtained in vivo. The Valsalva maneuver diminished the reflection from the aortic region of the renal arteries, thus allowing the distal reflected wave to become more evident on the thoracic pressure waveforms. Bilateral femoral artery occlusion usually enhanced the distal reflection and the Müller maneuver usually resulted in small increases in reflections. In conclusion, the geometric and elastic nonuniformity of the aorta results in two major sites of arterial wave reflection that influence the aortic pressure waveforms in man. The first major reflection site is located at the aortic level of the renal arterial branches and the second reflection site is located distal to the terminal aortic bifurcation.

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ALTERATIONS in the contour of the pressure wave along the aorta have been noted since the turn of the century, but the mechanisms responsible for these changes have not been fully explained. There have been few detailed investigations of the transmission characteristics of pressure along the human aorta. Studies by Luchsinger et al. and O’Rourke described and analyzed pressure waveforms in the human aorta, but pressures were not obtained simultaneously and a fluid-filled manometer system was used. Most of the previous laboratory investigations on this subject have been in animal preparations; however, it has been shown that the proximal aortic pulse contour in man differs from that of most laboratory animals, making it difficult to extrapolate all results in animals to man. Furthermore, quantitative assessment of changes in the arterial pressure waveform in man has been attempted in few laboratories because of the complex-
LATHAM et al.

ity in recording and analyzing multiple simultaneous high-fidelity arterial pressure signals.

Investigations performed in adult humans have shown that the proximal aortic pressure waveform commonly exhibits a secondary systolic wave (figure 1). This secondary systolic wave was generally considered to be associated with arterial wave reflections,1, 7, 8, 11, 13, 17 although there is strong theoretical and experimental support for the existence of arterial wave reflections,18 the site(s) of origin of these reflected waves is debated. Most often, arterial reflection sites have been ascribed to the region of the terminal aorta7, 9, 19 and/or the peripheral arteriolar bed.1, 5, 20

Arterial wave reflections are affected by local characteristic impedances that are directly proportional to pulse wave velocity and have been scrutinized frequently.21–24 Various methods have been used to obtain wave velocity,4, 20, 25–29 but due to the nonuniformities of the human aorta and technical limitations in its measurement, it is presently unclear which method best approximates true pulse wave velocity.

It is generally known that clinical maneuvers influence systemic arterial pressure, pulse wave velocity, and wave reflections.5, 8, 30 Previous studies have shown that the Valsalva maneuver reduces the secondary systolic wave in proximal aortic pressure waveforms (Figure 1) and makes the systemic input impedance spectrum less oscillatory, suggesting a diminution of reflections measured in the ascending aorta.8 An alternative explanation may be based on the observation by Kroeker and Wood,30 who reported a significant decrease in the duration of systole and a decrease in wave velocity during the Valsalva maneuver.3, 30 This may lead to a delay of reflected waves so that they arrive in diastole. Previous studies performed in our laboratory were confined to the ascending aorta so that the exact mechanisms of reduction of the secondary systolic wave were difficult to ascertain. Accordingly, this study was designed to investigate the entire human aorta and its terminal branches with simultaneous, equidistant (10 cm), micromanometric pressure measurements in order to determine regional pulse wave velocities, identify the major arterial reflection site(s), and assess the influence of clinical maneuvers on reflected waves. To aid in the interpretation of the results in vivo, a branching latex tube model was constructed based on the dimensions obtained from aortography.

Methods

Data acquisition. The patient population consisted of nine patients studied at Brooke Army Medical Center by elective cardiac catheterization for various chest pain syndromes. Patients included in this study were found to have normal coronary arteries and ventriculograms, as well as normal hemodynamic variables and ejection phase index values (table 1). Patients with hypertensive or clinical evidence of peripheral vascular disease were excluded.

This project was approved by the Clinical Investigation Committee and the Institutional Review Board at Brooke Army Medical Center. Informed consent was obtained for each study. All patients were studied by the right brachial artery approach. Cardiac outputs were obtained with a flow-directed thermodilution catheter positioned in the pulmonary artery. Pressures were initially obtained with a three-sensor catheter with transducers located at the tip, 4 cm proximal to the tip, and a third sensor 5 cm proximal to the middle one (Millar VPC-684H). This catheter was positioned with the tip in the cavity of the left ventricle, the middle sensor located just above the aortic valve, and the proximal sensor positioned in the aortic root. After a steady state was obtained and pressure waveforms were recorded in the control condition, the patient was asked to perform a standardized Valsalva maneuver.30 by exhaling into a mouthpiece connected to a one-way Rudolph valve. The effort expended was measured with a pressure gauge distal to the valve. A 40 mmHg expiratory effort was maintained for about 20 sec. The Müller maneuver was performed by reversing the valve and having the patient inhale with maximal inspiratory effort. Simultaneous femoral artery occlusions were then performed by bilateral manual compression at the level of the inguinal ligament. The patients rested until there was a return to control conditions between maneuvers.

The triple-sensor catheter was then withdrawn and replaced with a custom-designed catheter with six micromanometers mounted at 10 cm intervals [Millar PC-786(K)]. The catheter was advanced under fluoroscopic guidance until the tip sensor (No. 1) was in the distal external iliac artery and the most proximal sensor (No. 6) was positioned in the arch of the aorta (figure 2). All transducers were simultaneously calibrated with a precision laboratory mercury manometer before insertion.

Pressures from the six micromanometers were recorded under control conditions, during Valsalva and Müller maneuvers, and during bilateral femoral artery occlusion in the manner described above. The catheter position was documented by cinefluoroscopy and the distance between sensors was used to calibrate the measurements of aortic dimensions derived from the subsequent aortograms. Without altering the relationship between x-ray tube, patient, and image intensifier, the six-sensor catheter was withdrawn and then replaced with a No. 8F

FIGURE 1. Example of a pressure waveform from the ascending aorta of patient D. N. demonstrating a secondary systolic wave resulting from reflections. IP is the inflection point; Δtp is the transmission time of the reflected wave.
NIH angiographic catheter. This catheter was advanced to the aortic root, which was examined by single-plane aortography while approximately 100 ml of meglumine diatrizoate was injected at 20 ml/sec. Full-length cineangiograms were obtained by panning from the aortic arch to the femoral arteries.

**Data recording and analysis.** Data were recorded on FM analog tape (Honeywell 5600C) and converted offline to digital format at a sampling rate of 200 Hz after passing the signals through a low-pass filter (corner frequency 80 Hz, roll-off 12 dB/octave) to avoid aliasing. The foot of the pressure wave was determined by the intersection of slopes of the terminal diastolic and initial upstroke of the pressure wave at a paper speed of 200 mm/sec. Foot-to-peak wave velocity was measured as the time interval between intersection points of adjacent sensors. The time Atp was determined similarly. Fourier analysis was performed on individual beats to derive apparent phase velocity (see equation 4 below) and amplitudes of harmonics as a function of frequency.\(^{27,31}\) Mean apparent phase velocity was calculated by averaging values greater than 3 Hz.\(^{32}\) When the amplitude of pressure harmonics was less than 0.5 mm Hg, harmonics were considered to be too contaminated by noise to give useful information and discarded. Data from five to 10 cardiac cycles were averaged for each patient during control periods and interventions, where only plateau phases were analyzed. Harmonic amplitude changes along the aorta were calculated by taking the ratio of the pressure modulus at the specified location to the modulus of the same frequency of the most proximal pressure sensor (No. 6).

**Tube model.** With dimensions obtained from averaged radii of the aortograms, a latex model of the human descending aorta was constructed to compare the results of measurement of wave travel in vivo with results in the model. Dimensions were ob-

### TABLE 1

<table>
<thead>
<tr>
<th>Patient</th>
<th>Age (yr)</th>
<th>BSA (m²)</th>
<th>CO (l/min)</th>
<th>LVEDP (mm Hg)</th>
<th>LVEF (%)</th>
<th>HR (bpm)</th>
<th>MAP (mm Hg)</th>
<th>SVR (dynes-cm⁻²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>G. W.</td>
<td>49</td>
<td>2.12</td>
<td>7.1</td>
<td>10</td>
<td>77</td>
<td>68</td>
<td>88</td>
<td>1241</td>
</tr>
<tr>
<td>C. H.</td>
<td>35</td>
<td>1.88</td>
<td>5.8</td>
<td>7</td>
<td>57</td>
<td>76</td>
<td>90</td>
<td>945</td>
</tr>
<tr>
<td>P. S.</td>
<td>51</td>
<td>2.00</td>
<td>5.6</td>
<td>10</td>
<td>74</td>
<td>75</td>
<td>92</td>
<td>1271</td>
</tr>
<tr>
<td>A. B.</td>
<td>41</td>
<td>1.94</td>
<td>6.6</td>
<td>8</td>
<td>74</td>
<td>78</td>
<td>100</td>
<td>1333</td>
</tr>
<tr>
<td>J. I.</td>
<td>40</td>
<td>1.91</td>
<td>6.9</td>
<td>10</td>
<td>66</td>
<td>66</td>
<td>90</td>
<td>949</td>
</tr>
<tr>
<td>C. W.</td>
<td>37</td>
<td>2.20</td>
<td>6.4</td>
<td>9</td>
<td>79</td>
<td>71</td>
<td>95</td>
<td>1075</td>
</tr>
<tr>
<td>D. N.</td>
<td>42</td>
<td>1.92</td>
<td>8.1</td>
<td>12</td>
<td>74</td>
<td>58</td>
<td>105</td>
<td>1171</td>
</tr>
<tr>
<td>J. O.</td>
<td>39</td>
<td>2.00</td>
<td>5.4</td>
<td>10</td>
<td>68</td>
<td>63</td>
<td>87</td>
<td>1206</td>
</tr>
<tr>
<td>K. K.</td>
<td>40</td>
<td>1.78</td>
<td>6.3</td>
<td>7</td>
<td>84</td>
<td>69</td>
<td>75</td>
<td>899</td>
</tr>
<tr>
<td>Mean ± SD</td>
<td>42±5</td>
<td>1.97±0.12</td>
<td>6.5±0.8</td>
<td>9±1.6</td>
<td>73±8</td>
<td>69±6</td>
<td>91±8</td>
<td>1121±160</td>
</tr>
</tbody>
</table>

BSA = body surface area; CO = cardiac output; LVEDP = left ventricular end-diastolic pressure; LVEF = left ventricular ejection fraction; HR = heart rate; MAP = mean aortic root pressure; SVR = systemic vascular resistance.

\(^{a}\)K. K. is the only female patient.

**FIGURE 2.** Schema of the descending aorta identifying catheter sensor positions, indicated by numerals, and examples of pressure waveforms from patient C. H. Only one of the two ascending aortic pressure waveforms is depicted. The shaded area depicts the region of the aorta modeled by the elastic tube. The length of the tube model was constructed to accommodate the entire length of the six-sensor catheter. Diameter values ± SD are given.
TABLE 2
Model parameters

<table>
<thead>
<tr>
<th>Model</th>
<th>Internal diameter (cm)</th>
<th>Wall thickness (cm)</th>
<th>Length (cm)</th>
<th>$c_d$ (msec$^{-1}$)</th>
<th>$Z_c$ (dynes·sec·cm$^{-3}$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thoracic aorta</td>
<td>2.32</td>
<td>0.029</td>
<td>28.0</td>
<td>6.8</td>
<td>171</td>
</tr>
<tr>
<td>Abdominal aorta</td>
<td>1.40</td>
<td>0.039</td>
<td>25.0</td>
<td>8.8</td>
<td>606</td>
</tr>
<tr>
<td>Renal artery</td>
<td>0.96</td>
<td>0.026</td>
<td>8.8</td>
<td>11.1</td>
<td>1630</td>
</tr>
</tbody>
</table>

Dimensions of a latex tube constructed based on averaged values from aortograms of the patient population. All segments of the model are non tapering, and the elastic modulus of the material used was $15 \times 10^6$ dynes/cm$^2$.

$c_d$ = foot-to-foot wave velocity measured from impulse stimuli; $Z_c$ = characteristic impedance.

tained from averaged radii of the aortograms from the studies in vivo. The model consisted of a proximal uniform tube ("descending thoracic aorta") connected to a uniform distal tube ("abdominal aorta") and two branches ("renal arteries"). Dimensions for the model when mounted under 10% longitudinal stretch are given in table 2. The modulus of the elasticity for the material used was $15 \times 10^6$ dynes/cm$^2$ and the Moens-Korteweg equation (see equation 5 below) was used to adjust wall thickness to achieve pulse wave velocities resembling those measured in vivo. The resistances loading the "renal arteries" and the "abdominal aorta" were adjustable. Flow velocity was measured by a cannulated flow probe positioned at the proximal end of the model. Bench testing was performed with the use of sine waves and impulse stimuli. Pressure measurements were obtained with the same catheter [Millar PC-786(K)] used for in vivo measurements and sensors were placed in the same relative positions as in man (figure 2). The model differed from anatomy in vivo in that all three distal sensors were in a uniform tube referred to as the "abdominal aorta." No attempt was made to duplicate the terminal bifurcation or pelvic vessels, rather the resistor located at the terminus of the "abdominal" segment was used to represent total peripheral reflection. The tube system was filled with a 40% glycerin/60% NaCl mixture (0.9%) that had approximately the same viscosity (3.3 cP at 22°C) as blood at body temperature. Impulse stimuli were measured to obtain foot-to-foot wave velocity. Analog-to-digital conversion was performed on-line (sample rates of 500 Hz) with a PDP-11 Digital computer. The sine wave excitation was used to calculate apparent phase velocity (see equation 4). Wave velocities could be obtained for various settings of terminal resistances.

Statistical methods. A two-way analysis of variance (ANOVA) was used to test for significant differences between control and interventions. ANOVA also was used to test regional variances. We compared wave velocities of adjacent regions and control vs interventions. A paired t test was used to test for significant differences between foot-to-foot wave velocity and mean apparent wave velocity as well as differences in amplification of the first four harmonics between control and interventions. A value of p < .05 was accepted as indicative of statistic significance.

Theory. The time of the inflection point is the sum of the travel time of the forward wave and transmission time of the reflected wave, and may be determined from the foot of the pressure pulse to the inflection point, IP (figure 1). If the true pulse wave velocity ($c_d$) is known, then one may use the transmission time to calculate the distance (L) to the reflection site as follows:

$$L = c_d \times \Delta t/2$$  (1)

With the first minimum of the modulus of systemic input impedance ($f_{min}$) and the quarter-wave length formula, a second estimate of the distance to the reflection site or "effective length" (L) may be derived:

$$L = c_{ph}/4f_{min}$$  (2)

Regional foot-to-foot wave velocity ($c_{ph}$) was found by the measurement of the time interval (Δt) of the initial upstroke between two pressures a known distance (Δz) apart:

$$c_{ph} = \Delta z/\Delta t$$  (3)

Limitations to the technique arise from the difficulty in exactly determining the “foot” of the wave. Various methods have been proposed for measuring the foot of the wave and have been elucidated elsewhere.\textsuperscript{4,20,22} On the basis of experiments with uniform rubber tubes, it may be assumed that the foot of the wave is not affected by reflections and the foot-to-foot wave velocity therefore should approximate the $c_{ph}$.\textsuperscript{4,20}

Regional apparent phase velocity ($c_{app}$) was calculated over each interval (Δz = 10 cm) by subtracting the phase angles of corresponding harmonics (Δφ) at the two locations:

$$c_{app} = 2\pi n f_0 \Delta z/\Delta \phi_n$$  (4)

where f is heart rate, i.e., fundamental frequency and n is harmonic number.

Phase velocity may be estimated from the Moens-Korteweg equation,\textsuperscript{4,20}

$$c_{ph} = \sqrt{E h / 2 \pi R}$$  (5)

where E is the elastic (Young) modulus, h is the wall thickness, p is blood density, and R is the radius of the vessel. In a reflectionless system, the three velocities, $c_f$, mean $c_{app}$, and $c_{ph}$ should approach each other.\textsuperscript{33}

Reflections at arterial junctions are due to the mismatch in characteristic impedance on either side.\textsuperscript{5,23,24} The local reflection coefficient is calculated as:

$$\Gamma = \frac{1 - Z_r/Z_d}{1 + Z_r/Z_d}$$  (6)

where $Z_r$ is the characteristic impedance of the proximal tube and $Z_d$ is the characteristic impedance of the distal tube. The relationship between characteristic impedance ($Z_c$) and $c_{ph}$ is:

$$Z_c = p c_{ph} / A$$  (7)

where A is cross-sectional area. For parallel tubes (distal to bifurcations or trifurcations),\textsuperscript{4} impedances should be summed as parallel resistances. In equations 6 and 7 impedances used are real numbers, which is permitted for large arteries so that phases are zero and equations remain simple. Thus, one may calculate the reflection coefficient at an arterial junction when pulse wave velocities and areas are known, i.e., inserting equation 7 in equation 6.\textsuperscript{23} For a symmetrical bifurcation, the reflection coefficient is found to be:

$$\Gamma = \frac{1 - (2A_p c_p/A_p c_d)}{1 + (2A_p c_p/A_p c_d)}$$  (8)

where the proximal area and velocity are $A_p$ and $c_p$ and the distal parameters are $A_d$, $c_d$. For a trifurcation with two equal distal vessels, and a third vessel with different properties ($A_1$, $c_1$):
\[
\Gamma = \frac{1 - \frac{c_e}{A_p}}{1 + \frac{c_e}{A_p}} \left( \frac{A_1 + 2A_d}{c_1/c_d} \right)
\]

Results

Patient data are summarized in table 1. The population consisted of eight men and one woman with a mean age of 42 years. All patients were found to have normal coronary arteries and hemodynamic characteristics.

Aortic diameters were averaged at the aortic valve and sensor positions 5, 4, 3 and 2 (figure 2). Averaged diameters (±SD) were 2.6 ± 0.2, 2.3 ± 0.1, 2.0 ± 0.1, 1.3 ± 0.1, and 0.79 ± 0.03 cm, respectively. The diameter of the renal arteries was 0.78 ± 0.17 cm. These data revealed a significant taper (p < .0005) of the aorta, with the greatest change in diameter from proximal to distal of the origin of the renal arteries in the renal segment (34%).

During control, the proximal aortic pressure contour revealed an initial rise in pressure followed by an inflection point and a secondary systolic wave (figure 2). This inflection point occurred earlier (relative to the foot) in more distal waveforms until it was almost masked in the initial upstroke at location 4 (5 to 7 cm proximal to the renal arterial branches). The next sensor, No. 3, demonstrated another inflection point early in the waveshape, which again was obscured in the upstroke of the most distal sensor (No. 1). Measurements of the transmission time Δtp (figure 1) at the aortic valve, the ascending aorta (i.e., 5 cm distal, both from the triple tip sensor), and sensors 6, 5, and 4 from the six-sensor catheter were 140 ± 7, 120 ± 11, 107 ± 12, 71 ± 10, and 35 ± 9 msec, respectively. With the use of the average foot-to-foot velocity of the thoracic aorta (5.2 m/sec, see below), the distances calculated (equation 1) to the site of reflection were 36, 31, 28, 18, and 9 cm, respectively. The site of reflection determined by these distances was in the aortic segment at the level of the renal arterial branches. Furthermore, the total transmission time of the foot of the wave from the aortic valve to the renal artery branches was 68 ± 3 msec, which is almost exactly one-half Δtp of the waveform measured at the aortic valve (figure 1). Another inflection point at more distal sites was evident, with Δtp at sensor 2 being 27 ± 2 msec.

During the Müller maneuver, the foot of the reflected wave in the aortic root pressure occurred earlier than at control. The amplitude of the secondary wave was slightly increased (figure 3, A and B) in most cases. Bilateral femoral artery occlusion usually changed the character of the secondary systolic wave, which was increased in amplitude and more sharply peaked than at control in some patients (figure 3, A and C). During the plateau phase of the Valsalva maneuver, the most proximal aortic waveforms showed a loss of the secondary systolic wave (figure 3, A and D). At location 5 (10 cm distal to the arch), however, a small late systolic peak became manifest that increased in amplitude and occurred earlier in more distal waveforms.

The foot-to-foot wave velocity for each sensor pair is given in table 3. Each sensor pair defines a pulse wave velocity over a 10 cm segment of the aorta (5 cm in aortic root), with sensor pair 6–5 representing the most proximal portion of the descending aorta and sensor pair 2–1 located in the iliac artery segment. During control, there was regional variance in foot-to-foot wave velocity, with a significant increase (p < .01) at the region measured by sensor pair 3–2 from that at the adjacent proximal segment (sensor pair 4–3). During control, the Valsalva maneuver, and bilateral femoral artery occlusion, the velocities in the abdominal aorta were greater (p < .05) than the velocities in the thoracic aorta (not shown in table 3). During the Müller maneuver, proximal velocities were increased from control values (p < .05), resulting in a more uniform distribution of velocities throughout the aorta. Regional foot-to-foot velocities during the interventions were compared with their corresponding control values and significant differences are indicated in table 3.

The degree of amplification or attenuation of the individual harmonics with respect to location, averaged for all patients, is depicted in figure 4 for control and the three interventions. During control, amplitude of the fundamental harmonic tended to increase slightly from proximal to distal sites. The third and fourth harmonics increased significantly and plateaued by location 3. The rate of increase, however, was greatest between locations 5 and 4. During the Müller maneuver, little difference from control was noted in percent amplification of the first four harmonics, except at location 2 (p > .05). During the Valsalva maneuver, the first two harmonics continuously increased from proximal to distal sites; the relative increase being greater than during control (p < .005). The third and fourth harmonics, in contrast to control, exhibited an initial attenuation with an increase noted only in more distal sensors. Bilateral femoral artery occlusion produced an exaggerated degree of amplification in all
harmonics except the first, which was not significantly different from control (p > .05).

Regional apparent phase velocity was derived by applying equation 4. Mean apparent wave velocity as a function of location is shown in table 4 and is compared with foot-to-foot velocity in figure 5. It was found that during control the two wave velocities differed significantly (p < .01) at the region given by sensor pair 4–3. It is this region of the aorta in which the renal arteries have their origin and in which the aortic diameter significantly decreases (figure 2). The differences between the wave speeds were maintained during the Müller maneuver and femoral artery occlusion (p < .05 and p < .01, respectively), but disappeared during the Valsalva maneuver.

Apparent phase velocities as a function of frequency are presented in figure 6, together with the mean apparent phase velocities. Under control conditions apparent phase velocity in the low frequency range was found to be higher proximal to the renal arteries (sen-

<table>
<thead>
<tr>
<th>Condition</th>
<th>Aortic root</th>
<th>6-5</th>
<th>5-4</th>
<th>4-3</th>
<th>3-2</th>
<th>2-1</th>
</tr>
</thead>
<tbody>
<tr>
<td>Control</td>
<td>4.4 ± 0.4</td>
<td>5.3 ± 0.2</td>
<td>5.2 ± 0.1</td>
<td>5.7 ± 0.4</td>
<td>9.2 ± 0.5&lt;sup&gt;a&lt;/sup&gt;</td>
<td>8.8 ± 0.4</td>
</tr>
<tr>
<td>Valsalva</td>
<td>3.6 ± 0.2&lt;sup&gt;b&lt;/sup&gt;</td>
<td>5.1 ± 0.3</td>
<td>4.5 ± 0.3&lt;sup&gt;b&lt;/sup&gt;</td>
<td>4.5 ± 0.4&lt;sup&gt;b&lt;/sup&gt;</td>
<td>7.0 ± 0.7&lt;sup&gt;b&lt;/sup&gt;</td>
<td>7.5 ± 0.6&lt;sup&gt;b&lt;/sup&gt;</td>
</tr>
<tr>
<td>Müller</td>
<td>7.2 ± 0.1&lt;sup&gt;a&lt;/sup&gt;</td>
<td>6.6 ± 0.5&lt;sup&gt;a&lt;/sup&gt;</td>
<td>7.0 ± 1.2&lt;sup&gt;a&lt;/sup&gt;</td>
<td>5.5 ± 0.5</td>
<td>7.2 ± 1.2</td>
<td>7.3 ± 0.7</td>
</tr>
<tr>
<td>FAO</td>
<td>4.2 ± 0.8</td>
<td>5.5 ± 0.3</td>
<td>5.2 ± 0.2</td>
<td>5.3 ± 0.3&lt;sup&gt;b&lt;/sup&gt;</td>
<td>8.9 ± 0.7</td>
<td>9.5 ± 1.3</td>
</tr>
</tbody>
</table>

Values are means ± SEM. The aortic root velocity was measured with the triple-tip catheter with sensors 5 cm apart. The multiple-sensor catheter was used to determine velocity over 10 cm intervals. For locations, see figure 2.

FAO = femoral artery occlusion.

Comparison for each region is with its control value; comparison of control values is with adjacent regions: <sup>a</sup>p < .05; <sup>b</sup>p < .01.
sors 5, 4) than in the distal segment (sensors 3, 2). A minimum was reached at higher frequencies in the more proximal segments. The Valsalva maneuver attenuated the differences between proximal and distal segments. During bilateral femoral artery occlusion, the velocities at low frequencies in the proximal and distal aortic segments both deviated from the mean apparent wave velocity.

Reflection coefficients were calculated for the proximal descending aorta, the level of the renal arteries, and the region of the terminal bifurcation of the aorta with equations 6 to 9. Averaged aortic cross-sectional areas are given above and the foot-to-foot velocities are taken from table 3 (control). Pulse wave velocity in the renal arteries was assumed to be 1 m/sec greater than that in the abdominal segment. The local reflection coefficients were found to be small in the proximal descending aorta (from foot-to-foot velocities 6-5 and 5-4; cross-sectional area at locations A and B of figure 2, \( \Gamma = 0.05 \)) and at the terminal aortic bifurcation (from foot-to-foot velocities 3-2 and 2-1; cross-sectional areas of locations D and E of figure 2, \( \Gamma = 0.13 \)). The estimated characteristic impedances of the lower thoracic aorta, abdominal aorta, and renal arteries were 175, 730, and 2250 dynes-sec-cm\(^{-5} \), respectively. The reflection coefficient in the aorta at the level of the renal arterial branches was calculated to be 0.43.

**Model.** A tube model was constructed so that the effects of peripheral resistance and the mismatch around the renal arteries on reflections could be separated. The variability of apparent phase velocities, with frequency measured at sensor pair 3–2 (distal to “renal arteries”), demonstrated small oscillations around mean apparent phase velocity when peripheral reflections were small (figure 7). On the other hand, when the terminal resistor was completely closed, reflections were increased and the plot of apparent phase velocity was significantly altered (figure 7). Thus, with increased peripheral reflections, apparent phase velocity varies more with frequency, as demonstrated by previous investigators.\(^{30,34} \) More deviation of apparent phase velocity from its mean value was found in the model at sensor pair 5–4 (immediately proximal to the “renal arteries”) than in sensor pair 3–2 (distal to the “renal” branches), implying more reflections proximal to these branches (figure 8). In a similar fashion, the frequency variation of apparent phase velocity from sensor pair 5–4 in man indicated there was a stronger reflection there than in the abdominal aorta (figure 6). These data imply that the geometric config-

**FIGURE 4.** Amplification of the first four harmonics as a function of location during control, Valsalva and Müller maneuvers, and bilateral femoral artery occlusion averaged for all patients.

**TABLE 4**

<table>
<thead>
<tr>
<th>Condition</th>
<th>Sensor pair</th>
<th>Aortic root</th>
<th>6–5</th>
<th>5–4</th>
<th>4–3</th>
<th>3–2</th>
<th>2–1</th>
</tr>
</thead>
<tbody>
<tr>
<td>Control</td>
<td>4.3±0.2</td>
<td>5.4±0.5</td>
<td>5.3±0.3</td>
<td>10.4±0.9</td>
<td>8.8±0.7</td>
<td>6.4±0.4</td>
<td></td>
</tr>
<tr>
<td>Valsalva</td>
<td>4.0±0.3</td>
<td>4.8±0.3</td>
<td>4.0±0.2(^{a})</td>
<td>4.9±0.6(^{b})</td>
<td>7.7±0.6(^{a})</td>
<td>6.3±0.3</td>
<td></td>
</tr>
<tr>
<td>Müller</td>
<td>7.2±0.5(^{a})</td>
<td>6.4±1.2(^{a})</td>
<td>4.9±0.4</td>
<td>9.0±1.6</td>
<td>9.8±1.9</td>
<td>6.6±0.7</td>
<td></td>
</tr>
<tr>
<td>FAO</td>
<td>4.5±0.4</td>
<td>6.4±1.0</td>
<td>5.3±0.5</td>
<td>10.9±0.8</td>
<td>10.3±1.0</td>
<td>11.8±3.1</td>
<td></td>
</tr>
</tbody>
</table>

Values are ± SEM.
FAO = femoral artery occlusion.
Comparison for each region with control value: \(^{a}p < .05; \(^{b}p < .01.\)
uration of the descending aorta at the level of the renal arterial branches is a very likely site for the production of a major local reflection. Finally, the reflection coefficient could be precisely calculated in the model since foot-to-foot wave velocity in the "renal" segments was measured and tube diameters were determined by a micrometer. The renal reflection coefficient was found to be 0.34, which closely corresponded to our estimate in vivo of 0.43. The value of this model lies in its simplicity in testing the possibility of wave reflection from a particular geometric configuration analogous to the taper of the human aorta at the level of the renal arteries.

Discussion

This study is the first to employ simultaneous pressure measurements at multiple sites along the human aorta and to compare pressure wave transmission in a geometrically similar tube model to describe the effects of regional architecture on wave travel and wave reflection. The initial goal of our investigation was to identify the major site(s) of wave reflection in the aorta. To help identify these site(s), several characteristics of pulse wave propagation were analyzed: (1) inspection of regional waveform contours, (2) regional differences in wave velocity and pressure amplification, and (3) regional differences in the apparent phase velocity. Finally, a tube model of the human descending aorta was used to test the hypothesis that geometric nonuniformities in the aorta are capable of producing a reflected wave that is manifested in more proximal waveforms.

Luchsinger, O'Rourke, and Mills and their colleagues have reported pressure waveforms measured at different sites along the human aorta. Several mechanisms responsible for the differences in proximal and distal waveforms have been proposed, but the most widely accepted one is that of wave reflections. Reflection characteristics of the systemic circulation have been studied in a number of ways. McDonald and O'Rourke have proposed an "eccentric" or asymmetric T-tube model of the human circulation, implying that reflections are "effectively" summed at peripheral locations on opposite ends of the circulation. On the other hand, Murgo et al. concluded that, in addition to diffuse reflections from the periphery, the region of the terminal aortic bifurcation was a major reflection site. This conclusion is similar to those reached by other investigators. Both the inflection point on the aortic root waveform and the impedance spectrum were previously used to calculate the distance to a single dominant reflection site. With equa-
tions 1 and 2 and the pulse wave velocity measured in the aortic root, an average distance of 44 cm was calculated. One recognized limitation of this earlier study was the use of an aortic root wave velocity to estimate average wave velocity of the entire aorta. The use of the regional variation in pulse wave velocity could result in a length different from the distance to the reflection site derived earlier. The present study was undertaken to overcome these limitations and to analyze the mechanisms by which reflections are generated and transmitted in more detail.

The aortic root wave velocity in the present study is significantly different from that in the previous report. In that study a different catheter was used for the ascending aorta. The previously used catheter had only two sensors and when it was withdrawn from the left ventricle into the aortic root the native curve in the distal segment of the catheter resulted in a shorter axial distance between the two sensors. A special catheter, mounted with three micromanometers on which the two proximal transductors (5 cm apart) were mounted on a straight portion of the catheter, was used for the ascending aorta in our current study. This eliminated the possible error inherent in the previous study and resulted in pulse wave velocities we believe are more accurate and similar to previously reported values by King (1950), Barnett et al. (1961), and Merillon et al. (1978).

The waveforms recorded also were similar in contour to those previously reported. The inflection

FIGURE 6. Apparent phase velocity as a function of frequency at locations immediately proximal and distal to the renal branches during control, Valsalva and Müller maneuvers, and femoral artery occlusion. Data averaged (±SE) for all patients. Horizontal dashed lines represent mean apparent phase velocity.

FIGURE 7. Effect of reflections on apparent phase velocity as a function of frequency in the "abdominal" segment (sensor pair 3–2) of the model. Open circles refer to low resistance, i.e., little reflection, and closed circles represent high resistance, i.e., increased reflection.
point occurred earlier on the pressure waveform, proceeding from proximal to distal locations until it was manifest early on the initial upstroke at sensor 4, just above the renal arteries. A second inflection point, although less prominent, often was seen in more distal pressures and lost in the most distal waveform. Under control conditions, the time to the first inflection point and the foot-to-foot wave velocity from aortic root to renal arteries were used to calculate the site of reflection. This calculation showed the region of the aorta at about the level of the renal arterial branches to be a major reflection site. In support of this theory, the finding of a secondary systolic wave, as seen in aortic waveforms in man, was exactly reproduced by others by occluding the aorta of the dog at the level of the diaphragm. The second, more distal site was calculated to a distance slightly beyond the inguinal ligament. Wave speeds are high in the most distal segments and, although the reflected wave appears functionally discrete, it probably represents the summation of effects from the pelvic and lower extremity arteriolar beds, as previously described by O’Rourke. Therefore, we would suggest that the major local reflections in the human aorta can be described by two tubes of different properties in series, as proposed by Wetterer and Kenner. It is also possible there are other contributing regions to the total reflection, as proposed by Remington and O’Brien, which could not be evaluated in our study.

Although regional pulse wave velocity has been measured in animal preparations, there have been few analyses in man. It is very important when comparing results between laboratories that the criteria used to define the mean apparent phase velocity be specified. For the present study, mean apparent phase velocity represents the mean of values greater than 3 Hz. Both foot-to-foot and apparent wave velocity showed significant regional variation, although neither changed with location in a simple linear fashion. It is generally accepted that the wavefront used to measure foot-to-foot velocity is not affected by arterial reflections. When the two velocities are compared for each region, a significant difference is found in the aortic region at the level of the renal arteries, which can be interpreted to be the result of strong local reflections influencing the calculation of mean apparent phase velocity (figure 5). Thus, our data confirm that mean apparent phase velocity in arterial segments with strong local reflections should not be taken to be equivalent to foot-to-foot wave velocity or true phase velocity.

Furthermore, we found that the apparent phase velocity deviates more from its mean value with increased reflection, as shown by previous investigators. Under control conditions in vivo, the apparent phase velocity in the low frequency range was found to be higher in the aortic segment just proximal to the renal arteries than in that distal to the renal arteries.
The apparent phase velocity also reached a minimum at higher frequencies in the more proximal segment than in the distal segment. We conclude these findings to be consistent with a prominent local reflection in this region of the aorta.

The frequency of the first minimum of apparent phase velocity was not used to calculate the distance to the site of reflection ("effective length") by the quarter-wavelength formula (equation 2). The effective length calculated in this way is correct only for a uniform tube with frequency-independent load where the frequency of the first minimum of modulus of impedance \( f_{\text{min}} \) equals the frequency of the first crossover of the phase angle \( f_\phi \). This concept is questionable in a nonuniform tapering tube like the aorta. It has been shown in latex tubes and recently in vivo in our laboratory, that when \( f_{\text{min}} \) is not equal to \( f_\phi \), the calculated effective length significantly overestimates the actual distance to the site of reflection. Therefore, the frequency of the first minimum of the apparent phase velocity should only be used with caution to verify the distance to the site of reflection.

By means of Fourier analysis, pressure amplification was computed as well. Under control conditions, the greatest degree of amplification is in the region just proximal to the renal arterial branches, which also supports the conclusion of a prominent local reflection in the aortic segment incorporating the renal artery branches (figure 4). The maximal amplification of the third and fourth harmonic during the Valsalva maneuver and femoral occlusion differs considerably from control. This may be interpreted to mean that the aortic reflection from the renal artery level plays a lesser role during these interventions.

In addition to the above analyses, regional anatomic tapering of the aorta and its terminal branches was documented angiographically. The 10 cm segment of aorta measured by sensor pair 4–3 includes the region of transition of the thoracic aorta to the abdominal segment. It is also the section that passes through the diaphragm and incorporates the origin of the renal arteries. Inspection of aortograms from all patients revealed significant reduction in diameter of arteries distal to the renal arterial branches. This section contains the branching into the superior mesenteric artery, the coeliac artery, and two renal arteries and the continuing moderately sized abdominal aorta.

The importance of the discontinuity due to geometric and/or elastic factors at this segment is also supported by the calculated reflection coefficient that can be expressed in terms of the ratio of wave velocities and cross-sectional areas of proximal and distal segments of branching segments. We did not measure renal wave velocities in vivo, but if the region of the aorta at the level of the renal arteries were reflectionless, one would have to assume an unphysiologically low renal artery pulse wave velocity of about 1.5 m/sec. Assuming a renal artery pulse wave velocity approximately 1 m/sec greater than that of the abdominal aorta results in an estimated reflection coefficient of 0.43. A more detailed study of this aortic segment, including the effect of mesenteric and coeliac arteries, is necessary to fully understand reflections in this aortic segment.

Reported data on reflection coefficients in the aorta of man were obtained by determination of cross-sectional area in cadaver specimens and estimation of elastic wall properties. These studies arrived at a value of \( \Gamma = 0.16 \) for the terminal aortic bifurcation, and \( \Gamma = 0.25 \) at the level of the renal arteries. Our calculation of the reflection coefficients at the terminal aortic bifurcation and at the level of the renal arteries resulted in values of 0.13 and 0.43, respectively. The older data also indicate a relatively matched terminal aortic bifurcation compared with a rather large impedance mismatch of the aorta at the region of the renal artery branches.

The hydraulic model was chosen since other approaches (e.g., electrical and mathematical) are not able to incorporate all the detailed effects of bifurcations on pressure-flow relationships. The geometrically similar tube model was studied with the same multisensor catheter as was used for the determinations in vivo. From the tube model, we confirmed that apparent phase velocity was increased in the lower frequency range when the larger peripheral resistance was present (figure 7). The apparent phase velocity in the segment proximal to the "renal arteries" deviated more from its mean value than did the velocity in the distal segment (figure 8), consistent with a significant local reflection in the aortic region of the "renal" branches. If mean apparent phase velocity is calculated by averaging values larger than 3 Hz in the presence of strong local reflections, a value significantly different from the foot-to-foot velocity will be derived (figure 8 and table 2). Transmission times measured from impulse stimuli in the tube model also demonstrated a reflection from the geometric taper of the aorta at the region of the "renal" branches, which is consistent with findings in vivo.

Previous investigations have shown that clinical interventions in man may influence wave reflections. We have studied simultaneous pressures along the aorta during the physiologic stress induced by Valsalva and Müller maneuvers and bilateral fem-
oral artery occlusion. During the Valsalva maneuver, the increased intrathoracic pressure reduces the transmural pressure across the thoracic aortic wall significantly and decreases proximal aortic diameter by an average of 17%. The result is a thoracic aorta that is more compliant than during control. During the maneuver, foot-to-foot pulse wave velocity was significantly reduced throughout the aorta, with a relatively greater reduction of velocities in the abdominal segment. If the aortic diameters change with the same percentage, and the reduction in wave velocity is greater in the abdominal aorta than in the descending thoracic aorta (table 3), the renal junction becomes more matched (see equations 7, 8, and 9). Since this proximal reflection was reduced during Valsalva, reflection from a distal location, already seen in the control pressures, became more apparent.

During the Müller maneuver, there was a significant increase in pulse wave velocity in the proximal aorta to the extent that it approximated velocities of the distal segments (tables 3 and 4). The pressure waveforms during the Müller maneuver showed only slight increases in pulse pressure that mainly resulted from the increase in the amplitude of the secondary wave. Amplification (figure 4) was of a similar pattern compared with that under control conditions, indicating that reflection from the region of the renal artery branches was not further increased. The increase in uniformity of wave velocities throughout the aorta further implies that reflections are not necessarily increased. It may be speculated that the forward wave is larger during the Müller maneuver and returns earlier, leading to the observed pressure waveform. The normalized data in figure 4 do not account for different inputs.

In contrast, femoral artery occlusion by bilateral manual compression usually enhanced the reflected wave from the point of occlusion (figure 4). This was also manifested in proximal aortic pressure waveforms as an increase in the amplitude of the secondary systolic wave. The wave speeds in the proximal aorta were unchanged, and the time of the inflection point remained the same. Only the amplitude of the secondary wave was larger, thus indicating that this peak was the summation of reflection from both proximal (renal) and distal sites. The distal effective load is strongly frequency dependent, and the reflection coefficient may contain a significant phase angle, but it was not appropriate to calculate the location of this distal reflection site from the data presented here.

The present results do not reconcile all theories on arterial wave reflection in man, but provide a basis from which other studies may be performed to gain further insight into the hydrodynamic principles of the human circulation. A knowledge of wave propagation characteristics and vascular architecture gives the physician greater insight into the physiologic interpretation of wave shapes routinely measured in the proximal aorta. Reflection may be of importance for the cardiac workload (tension time index) and in future design of artificial grafts involving extensive segments of the human aorta.

We conclude that the physiologic stresses on the arterial system induced by clinical maneuvers produce alterations in regional wave reflections. The Valsalva maneuver diminishes reflection from the proximal site. The Müller maneuver affects reflection little, but equalizes wave speeds in the aorta, and femoral artery occlusion increases reflection from a distal site. We further conclude that there are two major sites of wave reflection from the lower part of the body that, along with the diffuse reflections from the periphery, affect aortic pressure waveforms in man. The first major reflection site appears to be at the aortic level of the renal artery branches and the second is distal to the terminal aortic bifurcation.

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